A Novel Volumetric Feature Extraction Technique with Applications to MR Images

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Abstract—A semiautomated feature extraction algorithm is presented for the extraction and measurement of the hippocampus from volumetric magnetic resonance imaging (MRI) head scans. This algorithm makes use of elements of both deformable model and region growing techniques and allows incorporation of *a priori* operator knowledge of hippocampal location and shape. Experimental results indicate that the algorithm is able to

hours or days. The model described in this paper requires no such tradeoff between resolution and computational intensity. More importantly, snake-based techniques do not incorporate any *a priori* model of the expected shape and size of the structure of interest. Therefore, they may not be useful for the identification of structures whose boundaries may be indistinct, such as the hippocampus, a gray-matter structure of the human brain, which is adjacent to other gray matter structures and has no distinguishable boundary along a significant portion of its surface. Our algorithm begins with a simply initialized shape model, composed of the superposition of multiple appropriately placed and shaped ovoids. This *a priori* modeling allows our algorithm to fill in areas of the surface of the structure of interest which have no apparent boundary in the data.

We propose a deformable model technique which incorporates some of the same goals as the region growing technique, which has been presented by Taylor and Barrett [12]. The Taylor algorithm provides competitive region growth from one or more seeds through comparison of border voxels to the first-order statistics of voxels which have already been absorbed. Our algorithm combines this concept with the idea of the deformable model. We begin with one or more seed voxels. Each of these seeds will, if left unconstrained, expand into an ovoid with a predetermined volume and preset ratios between radii in the x, y, and z directions. Constraining forces are elastic surface tension, deviation from the expected surface normal, and resistance from surrounding tissue. The expansive force is provided by internal pressure, which is gradually increased until either the expected volume is reached or no further expansion is possible due to constraining tissue.

One important application of this algorithm is in the quick and accurate *in vivo* volume measurement of the hippocampus and amygdala. Jack *et al.* and others [13], [14] have shown that such a measurement may be an important aid in the diagnosis of intractable temporal lobe epilepsy. Jack *et al.*

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actual series of MRI brain scans against a physician's manual identifications.

II. Modeling

The following deformable model algorithm is designed to operate on data which has previously been segmented by tissue type. An algorithm to provide such a segmentation for MR images has been presented previously in [2]. An algorithm for the segmentation of ultrasound images has been presented in [17].

As has been implied in Section I, this deformable model algorithm is essentially a region-growing technique, with added morphological constraints which come into play only under certain conditions. The physical system modeled is an expanding bubble with a preset geometry. The final volume of the model is limited by the ratio between internal pressure and elastic surface tension. Local surface morphology is controlled by the constraining force of surrounding tissue and by a penalty for deviation from the expected surface normal.

More precisely, the expansive force at a given boundary voxel is

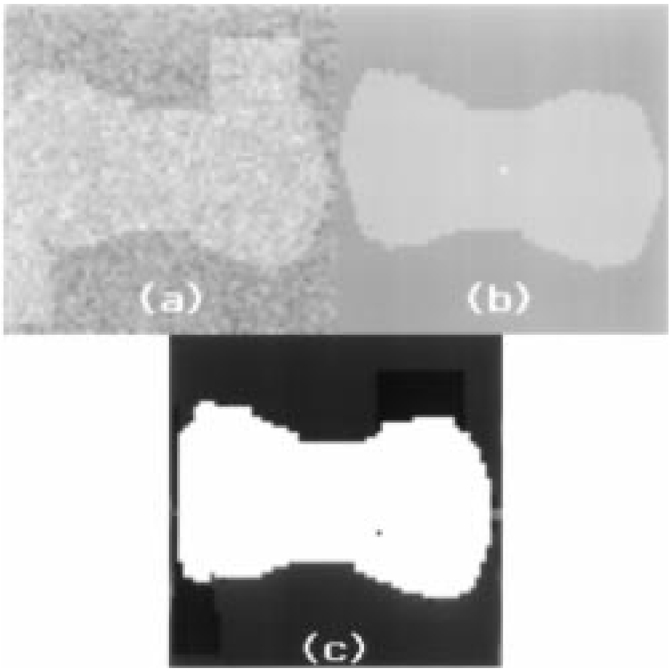
$$F = p - (S + N + C) \tag{1}$$

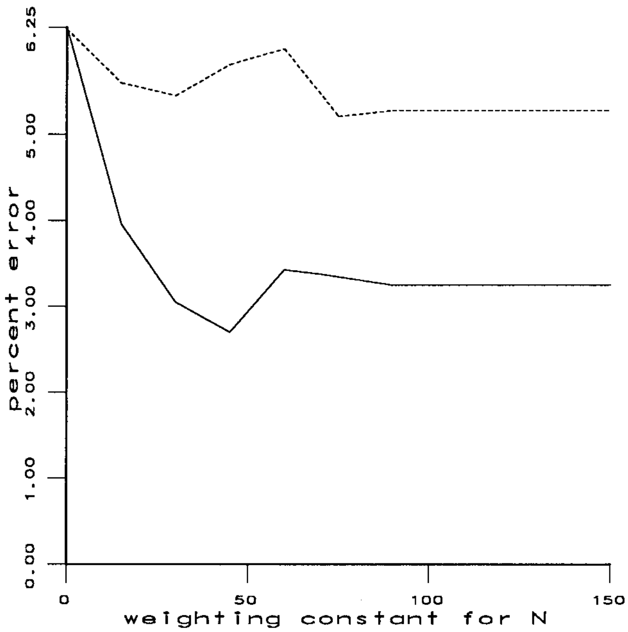
where p is internal pressure, given by

$$p = nRT/V \tag{2}$$

in which R is the universal gas constant, V is the current volume of the model, T is temperature, held constant at $1/R$, and n is set at the value necessary to reach force equilibrium at the expected final volume and surface area—i.e., the volume and surface area which the model would reach if it were allowed to grow unconstrained. This volume and area are calculated based upon an a priori shape model, the derivation of which will be discussed in the Section III. S in (1) is surface tension, which is proportional to the total surface area of the model. N is deviation from expected surface normal, given by

$$N = c_N \cos^{-1}(r_e \cdot r_a/(|r_e||r_a|)) \tag{3}$$


(a)
(b)
(c)



percent error
0.00
1.00
2.00
3.00
4.00
5.00
6.25
0
50
100
150
weighting constant for N

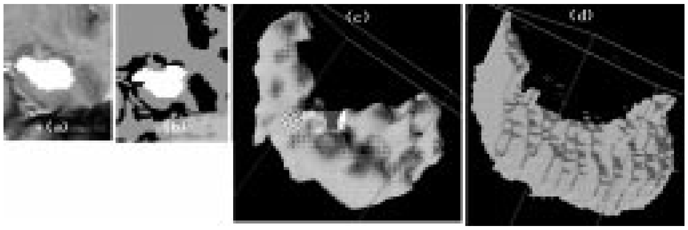


Fig. 10. (a) Physician's manual identi cation of the hippocampus on one slice of a coronal volumetric MRI head scan. (b) Hippocampus as identi ed by our algorithm on that same slice. (c) Volume-rendered reconstruction of the right hippocampus as identi ed by our algorithm. (d) Volume-rendered reconstruction of the right hippocampus as identi ed manually.

has reported a coef cient of variation in manual identi cation

TABLE I